Published in final edited form as:

IEEE Trans Nucl Sci. 2015 August ; 62(4): 1813–1824. doi:10.1109/TNS.2015.2446895.

Effect of Non-Alignment/Alignment of Attenuation Map Without/With Emission Motion Correction in Cardiac SPECT/CT

Joyoni Dey, W. Paul Segars, P. Hendrik Pretorius, and Michael A. King

Abstract

Purpose—We investigate the differences without/with respiratory motion correction in apparent imaging agent localization induced in reconstructed emission images when the attenuation maps used for attenuation correction (from CT) are misaligned with the patient anatomy during emission imaging due to differences in respiratory state.

Methods—We investigated use of attenuation maps acquired at different states of a 2 cm amplitude respiratory cycle (at end-expiration, at end-inspiration, the center map, the average transmission map, and a large breath-hold beyond range of respiration during emission imaging) to correct for attenuation in MLEM reconstruction for several anatomical variants of the NCAT phantom which included both with and without non-rigid motion between heart and sub-diaphragmatic regions (such as liver, kidneys etc). We tested these cases with and without emission motion correction and attenuation map alignment/non-alignment.

Results—For the NCAT default male anatomy the false count-reduction due to breathing was largely removed upon emission motion correction for the large majority of the cases. Exceptions (for the default male) were for the cases when using the large-breathhold end-inspiration map (TI_EXT), when we used the end-expiration (TE) map, and to a smaller extent, the end-inspiration map (TI). However moving the attenuation maps rigidly to align the heart region, reduced the remaining count-reduction artifacts. For the female patient count-reduction remained post motion correction using rigid map-alignment due to the breast soft-tissue misalignment. Quantitatively, after the transmission (rigid) alignment correction, the polar-map 17-segment RMS error with respect to the reference (motion-less case) reduced by 46.5% on average for the extreme breathhold case. The reductions were 40.8% for end-expiration map and 31.9% for end-inspiration cases on the average, comparable to the semi-ideal case where each state uses its own attenuation map for correction.

Conclusions—Two main conclusions are that even rigid emission motion correction to rigidly align the heart region to the attenuation map helps in average cases to reduce the count-reduction artifacts and secondly, within the limits of the study (ex. rigid correction) when there is lung tissue inferior to the heart as with the NCAT phantom employed in this study endexpiration maps (TE) might best be avoided as they may create more artifacts than the end-inspiration (TI) maps.

I. Introduction

Respiratory motion combined with the “respiratory creep” of the heart can result in ~15–20 mm motion of heart affecting the appearance of defects and thereby impacting the diagnostic accuracy of cardiac SPECT perfusion images [1]. In this work we investigate the differences in apparent imaging agent localization induced in emission images when the attenuation maps used for attenuation correction (from CT imaging) do not align with patient anatomy during emission imaging due to variations in respiratory state [2]–[13]. Recent papers have reported that the misalignment can cause severe to moderate defects in 40% of PET/CT cardiac studies [5] and 42% of SPECT/CT studies [3]. A major cause of this problem is differences in the position of the heart between breath-hold CT and free-breathing with emission imaging [6]. A currently favored method of correction is to acquire low mA CT studies with low pitch, slow rotation, and/or repeated rotation over a region (cine CT) to provide slices averaged over the respiratory cycle thus matching emission imaging [5–12]. However with sequential imaging there is still the possibility of drift in the respiratory pattern and body-motion which can cause alignment mismatches. Loghin et. al. [4] reported artifactual defects in 21.4% of their cardiac PET studies corrected by transmission imaging with a Ge-68 rotating rods source due to these reasons. Similarly, Gould et. al. [5] reported artifacts in 19% of their studies after correction by cine-CT, and Nye et. al. 2007 [12] reported artifacts in 28% of their studies AC with a slow-CT protocol. Thus manual or automatic registration of the attenuation map to the emission data is still needed to further reduce artifactual perfusion defects ([5], [13]).

In the work presented here, our first goal is to investigate (using the NCAT activity phantom and associated attenuation maps [16]) the extent of misalignment problem for different methods of procuring the attenuation map, when we did not attempt any correction for alignment of attenuation map to emission data. Particularly we want to see if there are inherently better configurations (such as end-inspiration versus end-expiration map) when procuring the breath-hold attenuation maps. We test on cases where there was rigid and non-rigid motion of heart between the heart and liver. Our second goal was to investigate to what extent, if any, emission and/or transmission motion correction based on just the 6-degree of freedom (DOF) motion of the heart region alone helps to compensate for these problems. The 6-DOF emission motion estimation and correction was performed using our prior algorithm [14]. Multi-modal registration to align the emission and transmission images is the subject of a whole investigation by itself (example [15]). In this study, the NCAT transmission attenuation maps are by construction aligned to one or the other of the emission motion states or by a known transformation to one of the motion states. Hence we simply used the known transformations to align the transmission and emissions.

II. Methods

A. SPECT Simulation of Rigid and Non-Rigid Motion with Irregular and Regular breathing

Different NCAT parameters were chosen to generate a variety of activity (and corresponding attenuation) maps for our investigation. Different angles of the heart were chosen to have different amount of lung tissue between the liver and heart, which we expect would affect the attenuation compensation if the maps are misaligned with the emission.

The NCAT angulation is designated as $zy_rotation$ (rotation about x-axis from +y-axis to +z axis), $xz_rotation$ (about y-axis from +z towards +x) and $yx_rotation$ (about z-axis, from +x towards +y). We chose three angular configurations: NCAT-default Male (DM) with the default heart-angle $(-90, -20, -50)^\circ$; Male (DA1) with a different heart angle such that the long axis is horizontal different heart-angle $(-90, 0, 0)^\circ$; and Male (DA2) with heart-angle $(-90, -40, -70)^\circ$. The DA1 would have more and DA2 would have less lung-tissue between the liver and heart than the default case.

Last, but not the least, the default female (DF) NCAT phantom case was considered. The breast parameters chosen were for default supine breasts with angulation tilted $\theta = 40.2$ deg transversely (sideways) from center of chest and $\phi = -20$ deg down. Long and short axes are 18.2cm and 7cm respectively. The NCAT size parameters are kept the same as the default male so as to primarily evaluate the effect of the breast soft-tissue.

Amplitude binning has proved more accurate than phase binning for respiratory motion correction of heart [17]. The standard NCAT software allows for outputs of respiratory states binned according to phase of the respiratory signal. Also the motion between the heart and liver is rigid, i.e. they move in unison. In [14] we used a modified version of NCAT code where the transformation of the heart (3 rotations and 3 translations) could be incorporated as parameters. Then the heart was kept at a single respiratory phase and linearly varying amplitudes were incorporated for the heart to generate given number of frames to achieve amplitude binning of the NCAT phantom. For this study the NCAT software was further modified (1) to allow for direct amplitude binning instead of phase-binning and (2) to easily allow for non-rigid motion between liver and heart. The non-rigid motion did not involve deformation, but was such that the heart and the liver each moved rigidly (with 6-DOF) but the motions could be set different from each other to within the constraints of the relative organs so as to avoid collision.

We set up rigid and non-rigid motion cases for our NCAT simulations. For the rigid case the amplitude of heart/liver motion was total of 2cm in SI and 0.6cm in AP. For the non-rigid case the heart motion was 2cm total in SI and 0.6 in AP, but the liver motion was 3cm in SI and 0.9cm in AP. The total amplitude of respiratory motion was divided evenly (linearly) within 18 amplitude states. Note that some non-rigidity during breathing will be inevitable for actual clinical cases; hence non-rigid motion was generated for all of the 4 anatomies (DM, DA1, DA2, DF). Rigid cases were generated just for the sake of comparison, for only the cases DM and DA1 which showed interesting results for non-rigid.

We used SIMIND to generate SPECT projections. The imaging agent simulated was Tc-99m sestamibi for cardiac perfusion SPECT imaging. As in [14] the relative concentration of activity in the NCAT phantom for the liver, gall bladder, kidneys, and spleen were one-half that of the heart, and that of the background was 1/10 that of the heart. The SIMIND acquisition mimicked a typical patient acquisition protocol considered in this study. A two-headed camera with low-energy-high-resolution (LEHR) collimators and a circular orbit of radius 28.36 cm was simulated. The collimators parameters used by SIMIND to model distance-dependent spatial resolution were a hole diameter of 1.22 mm, hole length of 27.0 mm, and a septal thickness of 0.20 mm. Classical scattering, Compton

scattering and photo-electric effect (accounting for attenuation) were included in the physics of the SIMIND acquisition. As detailed below, to maintain realistic counts, on nearly-noiseless acquisition via SIMIND the projections were appropriately scaled to clinical levels of 7.5million counts in total.

In more details, the 18 respiratory states were generated for all the projection angles with SIMIND with high (near-noiseless) count levels, and combined each 2 consecutive states to reduce to 9 sets of projections. The combining two-at-a-time makes sure there is some within-state motion, as would be expected clinically. We added noise so as to have 7.5 million counts over the 9 states, akin to an average patient study. We used an actual (irregular) patient breathing pattern and a uniform breathing pattern. With steady uniform breathing pattern the projections for each respiratory amplitude state are expected accumulate on average the same number of counts for a particular projection (i.e., with no preference for any particular state).

The irregular breathing pattern was chosen from one of the patient cases studied in [14]. As mentioned in [14] the patient was imaged with informed consent under an Institutional Review Board approved protocol. The acquisition was 2-headed with total of 68 angles. The time profiles of the two detector heads were identical as they were simultaneously acquired. The acquisition times for extreme respiratory States 1 and 9 are shown along with the reference position mid-state 5 in Figure 1(a). Figure 1(b) shows the total-times (summed over all angles) for each of the 9 respiratory states. The red-dashed line represents the time-duration if all the respiratory states had even distributions. For more details please refer to [14] (page 6463 Figure 13) where the acquisition time-distribution for each bin for all angles for this particular patient is plotted (in 0.1 secs). As explained in [14], the counts in the amplitude-binned SIMIND projections in each state were modified (modulated) by the true time distribution for the patient. Note that the SIMIND projections take into account attenuation variation with projection angle.

Irregular breathing was used for all anatomies - for non-rigid and (where applicable) for rigid cases. Uniform breathing applied only for DM non-rigid and rigid cases as a test. Thus we had $4+2+1+1=8$ cases to test: non-rigid and rigid DM with uniform and irregular motion (4 cases), non-rigid and rigid DA1 with irregular motion (2 cases), non-rigid DA2 (1 case) and finally the non-rigid DF (1 case). For each of these we had multiple attenuation map test cases (described in the next section) with and without emission motion correction.

All these datasets were to be compared against the corresponding motion-less case at the middle position. The projections at this state were scaled up to the full 7.5million counts. These projections were reconstructed to obtain the motion-less case at the reference middle position. This is called the Reference henceforward. This represents the case if the NCAT was motionless at the middle position. Just to recapitulate, we have 6 different references (motionless) cases -- 4 for the different non-rigid anatomies (DM, DA1, DA2, and DF) and 2 for the rigid anatomies (DM and DA1).

B. Attenuation Maps Investigated

For each of these datasets, we investigated various ways to correct for attenuation for SPECT/CT using the different NCAT emission maps and their corresponding attenuation maps. It is particularly noted, therefore, that the rigidity (non-rigid/rigid) was matched between the emission and transmission: that is, the non-rigid maps were used with the non-rigid emission cases while rigid attenuation maps were used with the rigid cases.

Two types of maps relating to the nature of respiratory motion during acquisition were investigated. The first type consisted of attenuation maps acquired at breath-hold at different points of respiratory cycle hence extents of respiration. These correspond to SPECT/CT acquired with a modern spiral CT where acquisition could be accomplished during a short enough period of time for the patient to be able to hold their breath. The breath-hold cases considered were breath-holds at normal end-inspiration and end-expiration and an extreme-breath-hold simulating the case where the patient takes a deep breath at the end of the inspiration. The second type was acquisition during free-breathing where various extents of respiration were combined at each acquisition angle. These could correspond for example to a “cine-CT” acquisition, a slowly rotating flat-panel system, or a spiral CT system slow rotation acquisition all during free-patient breathing. They could as well correspond with radionuclide-based transmission imaging which occurs simultaneous with emission imaging.

The NCAT activity and attenuation maps obtained for the 9 amplitude states (as explained in Section A) are denoted T1, T2, ... T9. T1 and T9 were respectively the end-inspiration and end-expiration maps and thus identified herein as TI (T1) and TE (T9). These were picked, and then combined or transformed as described below to simulate various SPECT/CT or transmission scenarios. The extreme breath-hold case was called TI-EXT. This was done for solely the default non-rigid male case (for irregular and uniform breathing), just to test the effect of the misalignment. In summary, the attenuation maps tested were:

1. Breath-hold SPECT/CT with CT-based attenuation maps acquired at breath-holds either at
 - a. end-inspiration: the **TI** (or T1)
 - b. end-expiration: the **TE** (or T9) or
 - c. an attenuation map for the reference: the T5 map, henceforth called **TR**. This may be clinically obtained approximately, by acquiring TI and TE (each at half-the-dose) and averaging the two extremes.
 - d. An extreme breath-hold case at end inspiration: the **TI-EXT**. This was generated by choosing 1.5 times the amplitude motion of the default case. Thus the TI_EXT would have about a 1.5cm displacement in SI direction from the reference position as opposed to the ± 1 cm displacement of TI or TE from reference.
 - e. TI moved rigidly to reference position (midway): the **TI_to_R**. This is done in order to match the heart position in the transmission scan with that of the emission scan at the mid-point of the breathing amplitude.

- f. Similarly, TE moved rigidly to reference position (midway): the **TE_to_R**
- g. TI-EXT moved rigidly to reference position (midway): the **TI_EXT_to_R** map.
- h. **Semi-ideal** case where all the 9 maps from T1, T2,..., T9 are used for the corresponding states. This was only semi-ideal (not ideal) for the non-rigid case because in our reconstruction we applied the motion of the heart rigidly. Thus for the case with non-rigid motion between liver and heart, there may be some residual mismatches.

As done typically in the literature, the CT-like attenuation maps were smoothed with a Gaussian with standard deviation of 5mm before applying them for attenuation correction in emission.

2. Average-CT with a attenuation map (called **Tave**), averaged over the respiratory cycles – such as if we did “cine-CT” or had transmission imaging by external sources.

C. Motion Estimation and Reconstruction

The methodology for emission motion correction used here is described in detail in [14]. Two passes through reconstruction were used using statistical Maximum Likelihood Expectation Maximization (MLEM) algorithm. In the first pass we reconstructed three-dimensional volume as done in [14] for the 9 amplitude-binned respiratory states with the goal of estimating 3D motion between them (described next). Reconstructions were done with collimation resolution compensation, scatter (TEW-based) and attenuation-map correction (using the average attenuation map of all positions).

From the reconstructed volumes for different respiration states we estimated the motion by registration relative to the middle state (which was used as the reference state). As explained in details in [14], for the case of irregular breathing, before reconstruction of each of the amplitude binned states the projection angles with sufficient counts to make material contributions to the reconstructed volume were determined for the given state, and the reference state. Only the binned projections whose angles were in common between the two sets were then employed in reconstruction. This was done to match the impact of limited angles in the reconstructed volumes of the two states being used to estimate the change in position of the heart during respiration.

The registration algorithm in [14] was used to estimate the 6-DOF rigid-body motion of the heart. An ellipsoidal volume around the heart in the reference state slices was segmented prior to registration so that it was the heart to which registration occurred and not the extra-cardiac activity concentrations, which can have different respiratory motion than the heart. Voxel intensity based registration was used to minimization of the sum-squared-difference of intensities between reference and motion-states to estimate the motion. Employing a large number of voxels in registration reduced susceptibility to noise. Based on our analysis in [14] with the level of noise considered in the current paper, the method is accurate to within 2mm.

In the second pass of reconstruction we used the emission motion estimates to perform motion correction and investigate different attenuation maps for correction. Thus, motion-correction, collimator resolution compensation, scatter-correction (TEW-based) and attenuation corrections were included within reconstruction but the key difference with [14] is that in this work, for the second pass reconstruction, the attenuation correction was performed with the various attenuation maps and methods to be investigated in this study.

The investigation of methodologies for multi-modal registration is out of scope for the current study. Therefore for the cases where transmission-map alignment to the emission motion corrected data (at reference position) was required we used the known values based on this being an NCAT simulation study. Thus when TE, TI, and TI_EXT needed to be moved to TE_R, TI_R, and TI_EXT_R respectively, the known transformations were used.

D. Polar-map Quantification

The results for the different cases were quantified by using polar maps on short-axis slices. All the different anatomies (eg DM, DA1, DF etc) were transformed independently for the short-axis view. However, for a given anatomy all cases were transformed the same way. As customary in the clinic, the variability due to noise across datasets is diminished by applying smoothing to the datasets before polar-map formation, matching parameters used when forming any database to which the polar map will be compared.

The polar-maps were divided into 17 segments as prescribed by the ASNC guidelines for reporting the results of SPECT myocardial perfusion imaging [19]. The 17 segment fractional uptakes values for each case were calculated and subtracted from corresponding reference segment to get an error value. The root mean square (RMS) of this error (averaged over 17 segments) was tabulated for all the cases. When appropriate the Maximum Difference (from the reference) of the 17 segments is reported.

III. Results

Figures 2–4 show representative polar maps for two of the male NCAT anatomies and the female anatomy. Figure 1 in the top row shows the polar maps for the default male with non-rigid irregular respiratory motion, for the interesting cases of TI, TI_to_R, TE, TE_to_R and TI_EXT and TI_EXT_to_R without correction of respiratory motion to mid-respiration during emission reconstruction. The bottom row shows the same cases with correction of respiratory motion in emission imaging. The left-most image in each row is the reference (motion-less case at the reference middle position). For the emission-motion-corrected bottom row, we see the improvement in going from TI to TI_to_R and from TE to TE_to_R. In fact for the motion corrected bottom row, we can state that the TI_to_R, TE_to_R and TI_EXT_to_R compare favorably to the reference, indicating that the rigid-motion correction of emission and transmission maybe adequate to correct for motion and attenuation map misalignment for the anatomy of this NCAT simulation.

Figure 3 shows the results for the DA1 NCAT anatomy in which the heart is horizontal in the chest and thus significantly impacted by the large superior / inferior component of respiratory motion.. The polar maps are scaled to their own maximum. Note that for this

case even the reference slices showed more non-uniformity than was seen for DM, and significantly more non-uniformity is seen for the extreme mismatches between CT imaging at breath-hold and emission imaging during breathing of TI and TE. There is some improvement in going from TE to TE_R in the bottom row (with emission motion correction).

Figure 4 shows the case of the female (DF) for which the case with emission motion correction (bottom row) retains severe count-reduction (defect-like) artifacts without and with attenuation map alignment due to non-rigid breast-tissue motion. Note the map-alignments are based on rigid heart motion. If there is significant soft-tissue over the chest area, like present for the female case, it would have significantly different motion from the heart, which would not be fully compensated in going from TI to TI_to_R or from TE to TE_to_R. Interestingly, in an apparent paradox, when the emission motion correction is not performed (top-row) the count reduction effect is lower (an example is pointed out with the white arrows) for all cases, whether the attenuation map is aligned or not. This is due to the blurring effect of respiratory motion. While it seems to apparently reduce the artifactual reduction in this case, this is not a desirable smoothing because the blurring due to the motion may conceal defects [14]. Furthermore, quantitative analyses on the uptake with respect to the reference, described next, points out that the Maximum or RMS difference from reference is in general smaller with emission motion correction than without it.

The quantitative analyses from polar maps are shown in Figures 5–6 and reported in Tables 2 and 3. In Figure 5 and 6 the first three subplots are the RMS difference of the uptake values of the 17 segments (from the corresponding values for the reference-case) for the key cases and anatomies. First three subplots of Figures 5 and 6 show the DM male non-rigid versus rigid (Figure 5) for irregular/uniform breathing; and with and without motion correction of irregular/uniform breathing for the male non-rigid case (Figure 6). From Figure 5 first three subplots we see that the non-rigidity had higher errors than the rigid cases for mis-aligned attenuation maps example TI and TE with the difference being more pronounced for TE. But the RMS errors for the non-rigid case were in general lowered to being comparable to the rigid cases for the instances where the attenuation maps were moved to reference position (ie. for TI_to_R or TE_to_R). The semi-ideal case was comparable for both. Figure 6 first three subplots show that in general the errors are higher for without emission correction and the improvement with correction is better for the cases for which the attenuation map was also moved to reference position.

The last subplot (titled DM vs DF) in Figures 5 and 6 shows comparisons of the Maximum Difference of the 17 segments (instead of RMS) from their respective reference for the non-rigid male and female cases. The maximum difference is more meaningful for comparison between the male versus female than the RMS. This is because the count reduction for the female over one or two segments is not well reflected in the RMS which is an average across all the 17 segments (as evident from a later table). Comparing the non-rigid male and female cases in Figure 5 we see that the maximum error is positive for males, and for the female case it is mostly negative reflecting the artifactual reduction in counts. For the one TE (female) case another segment other than the count-reduced segment happened to be dominant. In Figure 6 the last subplot shows without motion correction for the same non-

rigid DM and DF. The DM vs DF comparisons in Figures 5 and 6 clearly show the count reduction effect of the breasts of the female patient.

To further help in comparison of DM and DF the Maximum Difference is reported in in Table 2 for the non-rigid male and female cases without and with emission motion correction. Again we see that while for almost all the male-cases the maximum difference is positive, for the female case it is negative reflecting the artifactual reduction in counts. Note that semi-ideal case is for with-emission correction (with rigid motion estimations) and attenuation map alignment. Also note that in general the difference is less severe after emission motion correction than without for both DM and DF, confirming that the correction of the emission data to the midpoint between TI and TE decreases the impact of respiratory motion in attenuation correction.

The RMS difference for all the cases and anatomies are reported in Table 3 without and with emission motion correction. The last rows show the averages of the column-entries for without/with correction. With emission motion correction, the RMS error for the semi-ideal case expectedly was almost always lower than any other method, except TE_to_R. With motion correction, the average error across the 8 different NCATs for the semi-ideal case was 0.042. In corroboration with what was observed in the slices, the TE for non-rigid anatomies with motion correction was worse than the other attenuation map positions. The average RMS error for the TE-cases was 0.076 with emission motion correction. In contrast the average error with motion correction (last row) for TE_to_R (our simple rigid correction method) *was reduced by 40.8%* to 0.045 (to values comparable to the average RMS error of 0.042 of the semi-ideal). For the TI case, similarly the average error (with motion correction) was 0.069 and was reduced to 0.047 (again comparable to semi-ideal) for TI_to_R, i.e. a *31.9%* improvement. The most dramatic improvement was for the extreme breath-hold, where going from using the TI_EXT to TI_EXT_to_R reduced the average error from 0.114 to 0.061 or a *46.5%* improvement. Comparing non-rigid and rigid cases almost always showed more error for nonrigid. The one odd case was TE_to_R of rigid DM subject, with irregular breathing (with 0.072 RMS error for rigid as opposed to 0.045 for non-rigid), with emission motion correction. However the short axis slices do not show anything anomalous for TE_to_R rigid case and the visual differences across the rigid and non-rigid cases were very subtle.

The RMS errors for the *without* emission motion correction show that the improvements from aligning the attenuation map used for correction were more moderate if emission-motion correction was not performed. Moving the TI map to reference (TI_to_R) reduced the error from 0.081 to 0.069 on average, or a *14.8%* improvement. Moving the TE map to TE_to_R virtually kept the error unchanged (from 0.073 to just 0.072, or only a *1.4%* improvement). Moving the TI_EXT to reference (TI_EXT_to_R) did improve it, though not as much as with emission correction. The RMS error reduced from 0.115 to 0.078, or a *32.2%* improvement.

We observed that usage of a “cine-CT” or transmission imaging like method of acquiring the attenuation map (Tave) performed consistently as good as the best of the breath-hold

methods. Further improvement in cardiac uniformity with motion correction was likely due to improvements in heart uniformity with correction of respiratory motion.

Comparing without and with emission correction, the RMS of the fractional uptake values (see Section II D) are, in majority of cases, lower for with emission motion correction than without. The average error of all the entries for without-motion-correction cases was 0.079 while that for corresponding entries with-motion-correction was 0.064. Thus there was a overall reduction of error by 19%, on the average, with emission motion correction. For 7 cases (italicized in Table 3) the error indeed increased with emission correction. We reviewed each of these cases. Of the 7 cases, four were cases where remnant count-reduction was present for TE even after emission correction. But apart from that we did not find anomaly upon motion correction. For some cases, (such as shown with asterix in the example polar map set shown in Figure 2) a reason for increased error was due to the (normal) higher count-reduction in the basal septal sector region, where there is a shortening of the wall as part of the outflow-track of the left-ventricle (as shown with white arrow). Due to the respiratory induced smoothing this was less apparent for the uncorrected emission motion case.

The effects of attenuation map mismatch with emission were best appreciated on coronal slices due to the majority of the motion being in the superior / inferior direction. We show a representative example of our results in Figure 7, the case of irregular breathing for the default male (DM). The image at the left on the top row for Figure 7 is the reference (motionless case). The rest of the top row shows results without emission motion correction which show typically a moderate amount of respiratory induced count-reduction (green arrows) in both the superior and inferior walls. For the case of extreme breath-hold, TI_EXT, (highlighted with asterisk in Figure 7), the effect is severe in the superior wall which is adjacent to the lungs.

For most cases considered, these count-reduction effects were largely removed upon correction within reconstruction of the motion which occurred during emission imaging, as illustrated in second-row in Figure 7. The exception was for the TI-EXT, where the effect was still present (green-arrow) due to the large mismatch in attenuation between mid-respiration where the emission data was corrected to and the attenuation map acquired at exaggerated end-inspiration. Note that for the case where the attenuation map was moved rigidly to reference position by the known (by construction) transformation (TI_EXT_to_R highlighted with double asterisks), the count-reduction effect was reduced for *the emission motioncorrected case*.

IV. Discussion

In general it is observed here that with just 6-DOF motion corrections of emission *and* alignment correction of transmission, the degradation due to normal or even exaggerated breathing was largely removed. For some non-rigid cases, such as when using the TE map and (to a smaller extent) for the TI case, a small amount of count-reduction was still present in the inferior wall even after emission correction. Again, in the TE_to_R case we saw this effect was reduced.

The apparent superior and inferior wall count-reduction due to respiratory motion and its correction (by moving the map to the reference position) for TE, may be partially explained by studying the different attenuation maps. In Figure 8 we show the attenuation maps TE, TR and TE_to_R for the non-rigid male DM. We also show the absolute differences of the attenuation maps TE and TR, and those of TE_to_R and TR, to help illustrate the point. Notice the small sliver of lung-tissue under the heart (see arrows) would fall on part of the inferior wall of the emission motion corrected heart (which will be positioned similar to TR). This would result in an artifactual count-reduction in inferior wall during the iterative reconstruction which can be explained as follows. In the motion-corrected MLEM reconstruction process [18], consider the emission data at a particular iteration. This is positioned at reference (middle) position. It is then moved (rigidly in our algorithm) to a specific respiratory state and forward projected including attenuation, compared to the measurement at that state and the ratio is back-projected. When using TE map (heart at most superior location), the other 8 respiratory states (all other than the end-expiration state, or the 9th state) are misaligned to greater or smaller extent. This brings about the inferior wall count reduction. If we move the TE map to TR position (rigidly to align heart) and use this TE_to_R instead, this misalignment is reduced. The heart alignment of DA1 was horizontal in the chest which, even though an extreme case, would incorporate more lung tissue between the liver and the heart. Thus the artifactual count reduction is even more prominent for that case. Note for the rigid case the misalignment is smaller.

After emission motion correction the motion blur is reduced. However, we observed for some cases with misaligned attenuation maps the artifactual count reductions maybe more pronounced with respiratory motion correction during emission reconstruction than without motion correction. This is true particularly with *non-rigid* motion as illustrated in Figure 2. There an increased in non-uniformity can be seen when using the TE map, the extreme breathing TI_EXT, and also to a smaller extent for TI map. For the female patient this effect is seen for all cases as the misaligned soft-tissue in the chest for females may worsen the appearance of the polar-maps (Figure 2 and 4). Removing motion blurring in the emission slices for the heart can make this effect more pronounced. Motion blur is not a desirable effect however (and may hide/lower-contrast of lesions [14]), hence it is preferred to correct for both motion in the emission data and attenuation map misalignment. As evidenced in Figure 2, for the male patient in addition to emission motion-correction, after alignment of the attenuation map the appearance of the polar maps (TI_to_R, TE_to_R, and TI_EXT_to_R) is improved and more similar to the reference map. For the male NCAT this is also true for quantitative results in the two Tables 2 and 3. For the female NCAT however the chest-soft-tissue misalignment persists even after alignment of the heart-region and hence the false appearance of a defect persists for all cases. For females thus it is critical to perform non-rigid motion estimation to adequately align the attenuation-map soft-tissue to emission soft-tissue in the chest area.

For the case of the horizontal heart (DA1) there was still some residual count-reduction even after emission motion correction or transmission alignment via rigid movement of TE to the reference state. We believe this points to the weakness of using the end-expiration map (TE) in the first place compared to other positions when there is lung tissue inferior to the heart as

there was in our NCAT simulations (Figure 8). Also the effect of non-rigidity is more for this anatomy as evidenced in residual count-reduction even for the TI_to_R case. Thus for extreme/unusual anatomies non-rigid multimodal registration to estimate motion correction in emission and transmission seems desirable.

For TI or TE, we could also have chosen the 1st or the 9th state for reference – but the extreme states are often sparse in counts in the respiratory amplitude binned data [14]. A middle reference state is more robust for registration estimates. Thus moving the attenuation maps is one solution (as done here). Another solution may be to estimate motion via registration to a state with adequate counts (such as middle state) and transform the estimation parameters appropriately so as to reconstruct with a different position close to the breath-hold position where the attenuation maps were acquired (for example either of the end respiratory states).

Note as mention we did motion estimation via the emission datasets at different respiratory states and as appropriate applied them to transmission. This would work for NCAT, where the transmission maps such as TI, TE, TR etc and corresponding emission are perfectly aligned. However in reality the two modalities emission and CT are not acquired at the same time. Hence it is understood that for clinical cases registration of the heart region would be necessary for all cases for estimating the motion. Non-rigid multi-modal registration of the full torso is challenging because of missing anatomical data in SPECT compared to CT, uniformity of the liver region, and heart region in non-contrast CT studies used to form attenuation maps. This is being investigated in a separate work [15]. However, this paper shows even if a 6-DOF estimate is used to align the heart region with the attenuation map, we find reduction of artifacts for the average patient types.

The purpose of the study was to show the artifacts due to misalignment of transmission and emission that are present even for normal patients. Patients with ischemic defects or other abnormal heart-function may or may not be adversely affected or misdiagnosed if these abnormalities fall in the artifact region, depending on severity of the artifact and the defect. It should also be noted that some of our observations regarding respiratory motion correction might not be valid due to potentially higher inaccuracy in motion estimation for significantly abnormal patient studies. To fully investigate the impact of misalignment on abnormal studies would require a large systematic study of abnormal patients with different defect positions and severity and is out of scope of this paper.

V. Conclusions

From this study we concluded that the false count-reduction artifacts due to breathing were reduced for a large majority of cases with correction of respiratory motion to the mid-respiration reference point we used for the emission motion correction. Additionally that provided the transformation is known, even a simple rigid emission motion correction to rigidly align the heart region to the attenuation map helps in average cases to reduce the count-reduction artifacts.

Another finding of this paper is that in presence of the inevitable non-rigidity between the motion of the heart and liver, the end-expiration maps (TE) in individuals with lung tissue inferior to the heart might best be avoided as they may create more artifacts than the end-inspiration (TI) map. These observations are valid within the limits of the simulation studies performed in this experiment. We are further investigating 6DOF and non-rigid multi-modal motion estimates for NCAT, anthropomorphic and clinical studies.

Finally we observed that usage of a “cine-CT” or transmission imaging like method of acquiring the attenuation map (Tave) performed consistently as good as the best of the breath-hold methods. Further improvement in cardiac uniformity with motion correction was likely due to improvements in heart uniformity with correction of respiratory motion.

Acknowledgements

This work was supported by the National Institute of Biomedical Imaging and Bioengineering (NIBIB) grant R01 EB001457. The contents are solely the responsibility of the authors and do not necessarily represent the official views of the NIBIB.

Reference

1. Pretorius, PH.; King, MA.; Johnson, KL.; Mukherjee, JM.; Dey, J.; Konik, A. Combined Respiratory and Rigid Body Motion Compensation in Cardiac Perfusion SPECT Using a Visual Tracking System, 2011. IEEE NSS-MIC Conf. Record; Valencia; Spain. 2011 Oct. p. 2768-2773.2011
2. Le Meunier L, Maass-Moreno R, Carrasquillo JA, Dieckmann W, Bacharach SL. PET/CT imaging: Effect of respiratory motion on apparent myocardial uptake. J Nucl Cardiol. 2006; 13:821–830. [PubMed: 17174813]
3. Goetze S, Wahl RL. Prevalence of misregistration between SPECT and CT for attenuation-corrected myocardial perfusion SPECT. J Nucl Cardiol. 2007; 14:200–206. [PubMed: 17386382]
4. Loghin C, Sdringola S, Gould KL. Common artifacts in PET myocardial perfusion images due to attenuation-emission misregistration: clinical significance, causes, and solutions. J Nucl Med. 2004; 45:1029–1039. [PubMed: 15181138]
5. Gould KL, Pan TS, Loghin C, Johnson NP, Guha A, Sdringola S. Frequent diagnostic errors in cardiac PET/CT due to misregistration of CT attenuation and emission PET images: a definitive analysis of causes, consequences, and corrections. J Nucl Med. 2007; 48:1112–1121. [PubMed: 17574974]
6. Pan TS, Mawlawi O, Nehmeh SA, et al. Attenuation correction of PET images with respiration-averaged CT images in PET/CT. J Nucl Med. 2005; 46:1481–1487. [PubMed: 16157531]
7. Pan TS, Hamill JJ, Luo DS, Mawlawi O, Macapinlac HA. PET imaging of the thorax with attenuation correction by time-averaging and attenuation-averaging of CT data from 4DCT. J Nucl Med. 2006; 47:192P–193P.
8. Vesselle H, Kohlmyer S, Alessio A, Kinahan PE. Incorporation of cine CTAC to reduce the banana artifact in PET/CT imaging. J Nucl Med. 2006; 47:187P.
9. Alessio A, Kohlmyer S, Branch K, Chen G, Caldwell J, Kinahan PE. Cine CT for attenuation correction in cardiac PET/CT. J Nucl Med. 2007; 48:794–801. [PubMed: 17475969]
10. Chi P-C, Mawlawi O, Nehmeh SA, et al. Design of respiration averaged CT for attenuation correction of the PET data from PET/CT. Med Phys. 2007; 34:2039–2047. [PubMed: 17654907]
11. Cook RAH, Carnes G, Lee T-Y, Wells RG. Respiration-averaged CT for attenuation correction in canine cardiac PET/CT. J Nucl Med. 2007; 48:811–818. [PubMed: 17468434]
12. Nye JA, Esteves F, Votaw JR. Minimizing artifacts resulting from respiration and cardiac motion by optimization of the transmission scan in cardiac PET/CT. Med Phys. 2007; 34:1901–1906. [PubMed: 17654891]

13. Bond, SL.; Kadir, T.; Hamill, JJ.; Casey, ME.; Platsch, G. Automatic registration of cardiac PET-CT for attenuation correction; 2008 IEEE Medical Imaging Conference Record; 2008. p. M14-M16.
14. Dey J, Segars WP, Walvick RP, Bruyant PP, Pretorius PH, Dahlberg S, King MA. Estimation and Correction of Cardiac Respiratory Motion in Presence of Limited Angle Effects due to Irregular Respiration in SPECT. Med Phys. 2010 Dec; 37(12):6453–6465. [PubMed: 21302801]
15. Dey, J.; Konik, A.; King, MA. 2012 IEEE NSS-MIC. Anaheim, CA: 2012. Multi-modal Rigid and Non-Rigid Registration for Attenuation Correction in Cardiac SPECT/CT using Emission to CT conversion; p. 2859-2866.(oral)
16. Segars, WP. Biomedical Engineering. Chapel Hill, NC: University of North Carolina; 2001. Development and application of the new dynamic NURBS-based cardiac-torso (NCAT) phantom.
17. Lu W, Parikh PJ, Hubenschmidt JP, Bradley JD, Low DA. A comparison between amplitude sorting and phase-angle sorting using external respiratory measurement for 4D CT. Med. Phys. 2006 Aug; 33(8):2964–2974. [PubMed: 16964875]
18. Dey J, King MA. Theoretical and Numerical Study of MLEM and OSEM Reconstruction Algorithms for Motion Correction in Emission Tomography. IEEE Trans Nucl Sci. 2009 Oct 1; 56(5):2739–2749. [PubMed: 20419053]
19. Tilkemeier PL, Cooke CD, Grossman GB, McCallister D Jr, Ward RP. ASNC imaging guidelines for nuclear cardiology procedures. J. Nucl. Cardiol. 2009
20. Dey, J.; Segars, WP.; Pretorius, PH.; King, MA. Investigation of Respiratory Motion Correction with and without Motion Correction of Attenuation Map in Cardiac SPECT/CT in Presence of Non-Rigid Motion, an NCAT study; 2011 IEEE NSS-MIC Conf Record; 2011 Oct. p. 3237-3243.

(a)

(b)

Figure 1.

(a) Clinical acquisition times for a patient in msec for extreme-end respiratory states, 1 and 9 as well as the reference middle state 5. (b) The total acquisition time (sum over all angles) for each of the 9 respiratory states. The dashed line represents the hypothetical acquisition time if the respiration was uniform and the times were same for all the 9 respiratory states.

Figure 2.

Polar-maps from short-axis slices for Non-rigid Male (DM) with *irregular* breathing. (Top-Row): No emission respiratory motion correction. (Bottom-Row) With emission respiratory motion correction. The attenuation maps used in reconstruction starting are indicated above each polar-map and starting at the left the image in each row are the Reference, end-inspiration (TI), TI moved to reference (TI_to_R), end-expiration (TE), TE moved to reference (TE_to_R), TI_EXT (extreme-breath-hold) and TI_EXT moved to reference (TI_EXT_to_R). Note the basal-septal and apical count-reduction in the reference polar-maps is due to the anatomical thinning of the myocardial wall present in the NCAT phantom. The polar maps marked with asterisks indicate those characterized by having a higher ratio of RMS error with emission motion correction relative to their RMS error without motion correction. The arrow indicates higher count-reduction in the region of the outflow track of the left ventricle. The display color-bar is shown at the bottom.

Figure 3.

Polar-maps from short-axis slices for Non-rigid DA1 Male with horizontal angulation of the long-axis of the left ventricle with *irregular* breathing. (Top-Row): No emission motion correction. (Bottom-Row): With emission motion correction. As indicated above the maps starting at the left the image in each row is the Reference, TI, TI moved to reference (TI_to_R), TE, and TE moved to reference (TE_to_R). The display color-bar is shown at the bottom.

Figure 4.

Polar-maps from short-axis slices for non-rigid Female (DF) with *irregular* breathing. (Top-Row): No emission motion correction. (Bottom-Row) With emission motion correction. As indicated above the maps starting at the left the image in each row is the Reference, TI, TI moved to reference (TI_to_R), TE, TE moved to reference (TE_to_R). The arrows show example regions of apparent count reduction without or with motion correction, which due to removing motion blur, are accentuated with emission motion correction. The display color-bar is shown at the bottom.

Figure 5.

First three subplots (top two and left plot on bottom row) compare the *non-rigid vs. rigid* (with emission motion correction) comparisons for different male anatomies and breathing. They show the RMS error in polar-maps from reference for different attenuation map alignments to the emission data for the rigid and non-rigid motion for the DM (irregular and uniform breathing), and DA1 (irregular breathing) cases, respectively. The fourth subplot (with a different color scheme) shows the Maximum Difference (maximum difference from respective reference for the 17 segments) for *non-rigid male (DM) vs. female (DF)* for the irregular breathing case with motion correction. The respiratory motion simulated followed an irregular breathing pattern that was obtained from a patient study [14].

Figure 6.

First three subplots (top two and left plot on bottom row) compare the emission *motion correction* vs. *without motion correction* comparisons for different male anatomies and breathing. They show the RMS error in polar-maps from reference for different attenuation map alignments to the emission data for with and the without motion correction for the DM (irregular and uniform breathing), and DA1 (irregular breathing) cases, respectively. The fourth subplot (with a different color scheme) shows the Maximum Difference (maximum difference from respective reference for the 17 segments) for *non-rigid male (DM)* vs. *female (DF)* for the irregular breathing case, without motion correction. The respiratory motion simulated followed an irregular breathing pattern that was obtained from a patient study [14].

Figure 7.

Coronal Slices for Default Male (DM) NCAT anatomical geometry with *non-rigid, irregular, patient-like breathing*. (Top-Row) Reference is the left-most slice. All others are without emission motion correction and use the attenuation maps indicated in the figure. (Bottom-Row) shows coronal slices with emission motion correction reconstructed with the attenuation maps as indicated in the figure. The semi-ideal case is where the attenuation maps used for reconstructing each state matched that state (i.e., T1 for State1, T2 for State2...etc). Note the regions of false count-reduction induced by respiration present in the top row (no motion correction cases) as indicated by arrows are not apparent after emission motion correction (bottom row), except when using the TI_EXT map with the extreme breathing. This case shows significant count-reduction with/without motion correction. However the artifactual count reduction is reduced for case TI_EXT_to_R where the map is (rigidly) moved to reference position.

Figure 8.

Coronal slices of original attenuation maps of NCAT through the mid-level of the heart for Male (DM). From left to right, attenuation maps TE, TR, TE_to_R and the Absolute|TE-TR| and Absolute|TE_to_R - TR|. Notice the lung-tissue under the heart (see arrow) would fall where the inferior wall of emission motion corrected heart (positioned similar to TR) would fall. This may cause a small count-reduction artifact in the inferior wall. We observe that if we move the TE map to TR position (even rigidly to at least align the heart) this effect is reduced or removed.

Table 2

Polar Map quantification without and corresponding with emission motion correction: **Maximum difference** of 17 segments with respect to corresponding reference (unit-less)

	Motion Corr	Semi-Ideal	TI	TE	Tave	TR	TI_to_R	TE_to_R
Male (DM), Non-Rigid Irregular-Breathing	Without	--	0.209	0.192	0.175	0.185	0.174	0.191
	With	0.085	0.155	0.159	0.122	0.120	0.107	0.088
Female (DF), Non-Rigid, Irregular-Breathing	Without	--	-0.104	0.134	-0.113	-0.105	-0.113	-0.128
	With	-0.078	-0.087	0.134	-0.068	-0.060	-0.080	-0.075

Table 3

Polar Map quantification without and corresponding with emission motion correction: **RMS error** of fractional uptake values for 17 segments with respect to corresponding reference (unit-less). *

	Motion Corr	Semi-Ideal	TI	TE	Tave	TR	TI_to_R	TE_to_R	TI_EXT	TI_EXT_to_R
Male (DM), Non-Rigid Irregular-Breathing	Without	--	0.092	0.067	0.074	0.078	0.077	0.078	0.112	0.080
	With	0.048	0.067	0.110	0.062	0.064	0.051	0.045	0.121	0.065
Male (DM), Rigid, Irregular-Breathing	Without	--	0.097	0.059	0.061	0.065	0.064	0.064	--	--
	With	0.026	0.065	0.069	0.037	0.034	0.056	0.072	--	--
Male (DM), Non-Rigid Uniform-Breathing	Without	--	0.083	0.077	0.066	0.069	0.072	0.070	0.118	0.076
	With	0.041	0.082	0.084	0.044	0.043	0.041	0.031	0.108	0.056
Male (DM), Rigid, Uniform-Breathing	Without	--	0.081	0.057	0.049	0.052	0.055	0.051	--	--
	With	0.031	0.065	0.065	0.047	0.045	0.033	0.027	--	--
Male (DA1), Non-Rigid Irregular-Breathing	Without	--	0.092	0.110	0.093	0.092	0.088	0.098	--	--
	With	0.048	0.086	0.101	0.053	0.050	0.042	0.057	--	--
Male (DA1), Rigid Irregular-Breathing	Without	--	0.068	0.066	0.063	0.063	0.062	0.065	--	--
	With	0.038	0.057	0.043	0.040	0.040	0.037	0.037	--	--
Male (DA2), Non-Rigid, Irregular-Breathing	Without	--	0.081	0.069	0.074	0.076	0.078	0.069	--	--
	With	0.060	0.087	0.066	0.073	0.075	0.074	0.050	--	--
Female (DF), Non-Rigid Irregular-Breathing	Without	--	0.057	0.077	0.073	0.071	0.058	0.082	--	--
	With	0.042	0.047	0.068	0.036	0.035	0.043	0.039	--	--
Average	Without	--	0.081	0.073	0.069	0.071	0.069	0.072	0.115	0.078
	With	0.042	0.069	0.076	0.049	0.048	0.047	0.045	0.114	0.061

* Note: The italicized entries are cases where the fractional uptake RMS error values from reference are less for without emission motion correction than with the correction